Contrast Matching Techniques for Digital Subtraction Radiography: An Objective Evaluation

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Digital subtraction radiography (DSR) enables the detection of subtle early detrimental effects of periodontal disease as well as the evaluation of the effects of therapy. However, the differences between two radiographs due to alignment and contrast errors must be kept at minimum. In the present in vitro study we test the efficacy of three basic contrast correction methods in the reduction of contrast mismatches which can adversely affect a subtracted image. The ODTF (Optical Density Thickness Function) method, which is based on a function relating grey level values of the aluminium wedge image and the corresponding thicknesses of the wedge, induced less contrast correction error than the CDF (Cumulative Density Function) and the LSQA (Least Square Quadratic Approximation) methods. Moreover, CDF, ODTF, and LSQA functions obtained from the reference structure density distribution may be applied for objective contrast enhancements and for standardisation of image quality, while the ODTF function allows also bone change volume estimations.

INTRODUCTION

Digital subtraction radiography (DSR) is a powerful diagnostic tool for the interpretation of periodontal processes in longitudinal studies.[1] It is performed by superimposing and subtracting a pair of radiographs, one obtained at the beginning and the other at the end of a time period of interest. If both radiographs are obtained under identical conditions as projection geometry, x-ray tube settings, and film developing procedures, the subtracted image will show only the structures that have changed. By this technique even subtle changes in bone or tooth tissue, not perceptible by direct comparison of films, can be revealed. Furthermore, by using reference structures and by relating thicknesses and densities, DSR can be used to quantify the changes, not only to visualise them.[2,3] Because the conditions are very hard to maintain constant over time, it is necessary to develop suitable procedures to reduce or even eliminate the differences induced by uneven conditions. Various methods have been proposed to correct the projective distortions in dental radiographs.[4,5] Although perfectly aligned, the fluctuations in x-ray tube settings and film developing procedures will result in different contrast between the two radiographs. The subtracted image will show actual changes as well as contrast errors and can therefore be misinterpreted. Many methods have been applied to eliminate contrast errors,[6-9] but little information is available on quantitative effects of such corrections.[9]
The aim of this study was to test the efficacy of three contrast correction methods[6-8] in the reduction of contrast mismatches without corrupting real changes between two radiographic images.

MATERIALS AND METHODS

Image acquisition

Five radiographs of the same slice of a mandible of a beagle dog with an included reference aluminium wedge (29mm x 16mm x 5mm) were radiographed (PHILIPS long tube, 65kV, 7.5mA) on AGFA DENTUS M2 film (size 2, film speed D) at different exposure times 0.14, 0.18, 0.22, 0.26, and 0.30 seconds, respectively. The radiographs were placed on the negatoscope and the images were captured by a monochrome CCD (charge-coupled device) video camera (Sony AVC-D7CE, Japan) and digitised by a frame grabber (VFG Visionetics, Taiwan) as 256x256x8bit images. The five images were used as reference images. Five copies of each digital reference image were made. In each copy bone loss was simulated by reducing the grey level value of each pixel by 10 in a region representing 1, 5, 10, 15 or 20% of the whole image.

Contrast correction methods

CDF contrast correction method. The method proposed by Ruttiman et al.[6] is based on close matching of cumulative density functions (CDF) of identical reference regions in two radiographs.

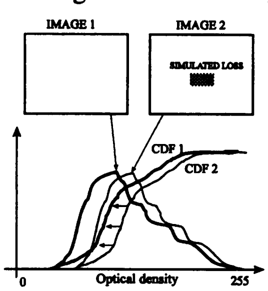

Figure 1: CDF method

Grey levels of all pixels, even those representing simulated bone loss, were included in calculations of CDFs. The grey level mapping procedure is

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performed by pooling or shifting bins associated with the histogram to be modified in such a way that the cumulative sum is matched as closely as possible to the corresponding sum in the target histogram, without splitting existing bins into different grey levels (Figure 1).

ODTF contrast correction method. The optical density thickness function (ODTF) is a function that gives the relationship between each grey level value in the image representing the aluminium wedge and the corresponding thickness of the wedge (Figure 2). It is a modified version of the aluminium equivalent thickness (AET) function proposed by Vos et al.[7]

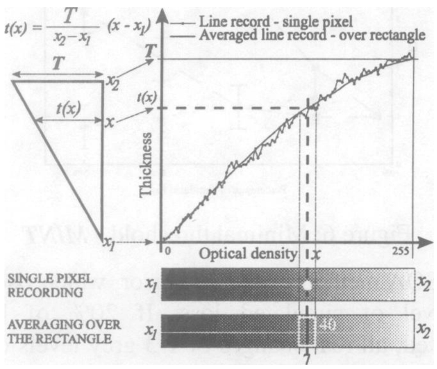

Figure 2: ODTF recording

To obtain the ODTF the co-ordinates of the leftmost (x_1) and the rightmost (x_2) pixel in the middle of the image of the Al-wedge are defined. For each pixel having the position x on the line connecting x_1 to x_2, the optical density *od(x)* is obtained by averaging the optical densities of 7x40 pixels forming a rectangular region and its corresponding thickness *t(x)* is calculated as $t(x)=T*(x-x_1)/(x_2-x_1)$ (Figure 2).

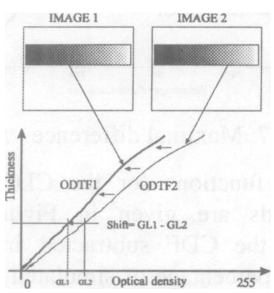

Figure 3: ODTF method

The contrast correction of an image with respect to a reference image is based on recording the optical density thickness functions of both images and closely matching them (Figure 3).

LSQA contrast correction method. The least squares quadratic approximation (LSQA) contrast correction method requires that two images are aligned before they are contrast corrected.[8] The grey levels of all pairs of pixels having the same spatial co-ordinates in the first and the second image are recorded in the 2-dimensional space and a regression curve, which is a quadratic polynomial obtained by least squares approximation, is used as a mapping function between the grey level values of two images (Figure 4).

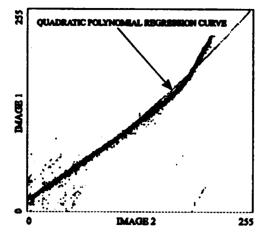

Figure 4: LSQA method

Contrast corrections and subtraction of the radiographs. Each copy was contrast corrected by the CDF, ODTF, and the LSQA method and subtracted from its original image. This resulted in 75 subtracted images, 25 for ODTF, 25 for CDF, and 25 for LSQA, respectively. A fixed grey level value of 127 was added to the value of each pixel in the subtracted image to produce an image in which grey level values below 127 represented loss and values above 127 remodelling.

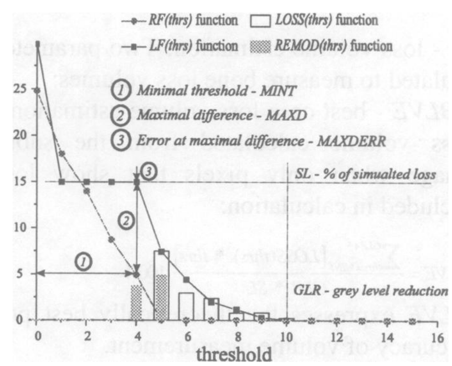

Figure 5: Testing functions and three parameters shown for the case of 15% of simulated loss

Parameters for the evaluation of contrast correction methods

In order to test the potential of each method to correct grey level values between two images subtracted images were automatically thresholded by setting the threshold (*thrs*) from 0 to 15, step 1. Pixels with grey level values greater than (*127+thrs*) were treated as pixels that show bone remodelling and those, which had grey level values less than (*127-thrs*) were treated as bone loss. All other pixels were treated as pixels that show no change in tissue. To express the extent of revealed bone loss (remodelling) at each threshold two functions of the threshold were formed, the Loss Function (*LF(thrs)*) and the Remodelling Function (*RF(thrs)*) (Figure 5). Additional two testing functions, *LOSS(thrs)* and *REMOD(thrs)*, express the proportions of pixels that have a grey level value equal to (*127-thrs*) and to (*127+thrs*), respectively.

Using the testing functions, the simulated loss (*SL* = 1, 5, 10, 15 or 20%), and the value of grey level reduction (*GLR*=10) four parameters were defined:

1. *MINT* - minimal threshold necessary to eliminate most of the contrast correction error. *MINT* is the smallest threshold at which *LF(thrs)* falls under *3/2* of simulated loss and *RF(thrs)* falls under *1/2* of simulated loss.
2. *MAXD* - maximal difference between *LF(thrs)* and *RF(thrs)*:

$$MAXD = \max_{\substack{thrs \\ thrs >= MINT}} \frac{LF(thrs) - RF(thrs)}{SL} 100\% \tag{1}$$

Ideally, *MAXD* would be 100%.

3. *MAXDERR* - error of loss measurement at threshold $thrs^*$ at which *MAXD* was found:

$$MAXDERR = \frac{|IL - LF(thrs^*)|}{SL} 100\% \tag{2}$$

MAXDERR is the absolute difference between simulated and revealed loss at $thrs^*$. It is an indicator of how accurate bone loss regions can be detected.

4. *LVE* - loss volume estimation. Two parameters are calculated to measure bone loss volumes:

4a. *BLVE* - best case loss volume estimation is the loss volume calculated from the subtracted image when only pixels that show loss are included in calculation:

$$BLVE = \frac{\sum_{thrs=MINT}^{GLR+5} [LOSS(thrs) * thrs]}{GLR * SL} 100\% \tag{3}$$

BLVE expresses the theoretically best possible accuracy of volume measurement.

4b. *WLVE* - worst case loss volume estimation is the estimation of bone loss volume when pixels that show loss as well as remodelling are included in the calculation:

$$WLVE = \frac{\sum_{thrs=MINT}^{GLR+5} [(LOSS(thrs) - REMOD(thrs)) * thrs]}{GLR * SL} 100\% \tag{4}$$

BWLE and *WLVE* are expressed in percentage of simulated bone loss volume, defined as (*GLR*SL*), and are ideally 100%.

Each subtracted image was evaluated by all four parameters. The data are expressed as means ±standard deviations (*n*=5) and plotted as a function of simulated loss for all three contrast correction methods.

RESULTS

Minimal threshold *MINT* necessary to eliminate most of the contrast correction error is plotted as a function of simulated loss in Figure 6. For 1% of simulated loss a threshold of 1 can be applied in all methods. Therefore, only pixels that have changed for only one grey level may not be classified with certainty as loss or gain. If an image is contrast corrected by the ODTF method a low threshold of 1 can be used for all levels of simulated loss.

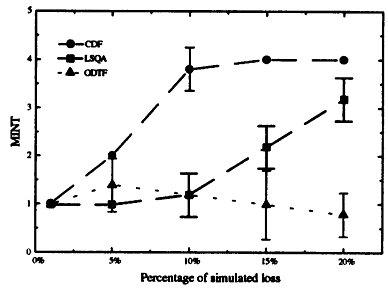

Figure 6: Minimal threshold - *MINT*

The LSQA method adds more error, which rises with the level of simulated loss. If 20% of loss is simulated, all real changes of 1-3 grey levels can not be revealed. The CDF method adds the largest amount of error to the subtracted image and therefore changes of 1-4 grey levels can not be detected if 10% or more loss is simulated.

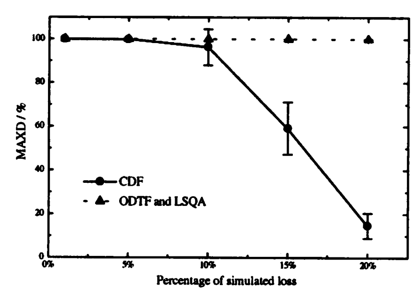

Figure 7: Maximal difference - *MAXD*

The *MAXD* functions for the CDF, ODTF and LSQA methods are given in Figure 7. *MAXD* measured on the CDF subtracted image declines rapidly as the percentage of simulated loss increases from 10% to 20%, while for low percentage of simulated loss, *MAXD* is close to ideal 100%. If 15% of bone loss is simulated, only 60% of it can be revealed. *MAXD* for the ODTF and LSQA methods is not correlated with the percentage of the simulated loss and is constantly 100%. Consequently, real loss regions can be totally revealed from the subtracted image. *MAXDERR* is plotted in Figure 8 for all three methods. ODTF and LSQA methods show no error for all levels of simulated loss. *MAXDERR* for the

CDF method increases with the amount of simulated loss and reaches almost 90% at 20% of simulated loss.

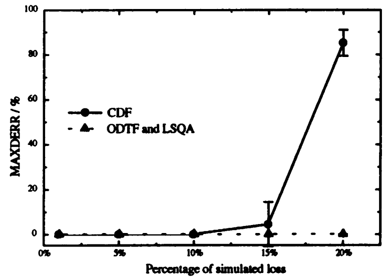

Figure 8: Error at maximal difference - MAXDERR

Bone Loss Volume Estimation *LVE* for the CDF method is presented in Figure 9. The expected value of *LVE* is somewhere between best possible estimation *BLVE* and worst possible estimation *WLVE*. It is obvious that *LVE* is strongly related with the level of simulated loss. Only at real small changes in tissue density (1% of pixels with grey level reduced by 10 or about 10% of bone thickness) a 100% accuracy can be expected. At 5% of simulated loss only 85% of real loss volume can be retrieved from the subtracted image, at 10% only about 70%, at 15% from 30 to 60%, and at 20% of simulated loss it is almost impossible to reveal any bone loss from the subtracted image. Hence, the CDF contrast correction method tends to obscure actual changes between two images.

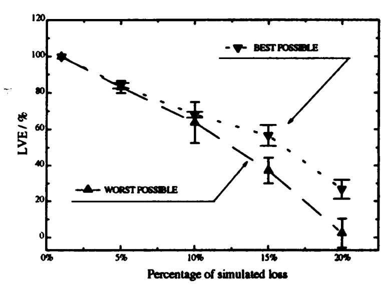

Figure 9: Loss volume estimation - *LVE* for the CDF subtracted images

BLVE equals *WLVE* in ODTF and LSQA methods because all pixels that show remodelling can be successfully eliminated from calculations by applying a small threshold. When using the ODTF method *LVE* is not correlated with the level of simulated loss (Figure 10). Accuracy of the *LVE* for the LSQA method (Figure 10) is correlated with the amount of the simulated loss. Although this method is more accurate than the CDF, only 70% of real loss volume can be revealed from the LSQA subtracted images at 20% of simulated loss.

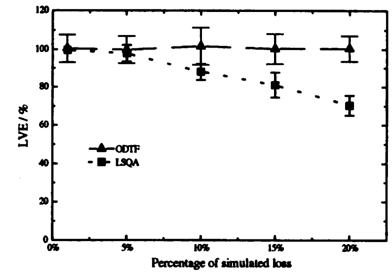

Figure 10: Loss volume estimation - *LVE* for the ODTF and LSQA subtracted images

DISCUSSION AND CONCLUSION

It is well known that each contrast correction method induces some error, which in the subtracted image manifests as bone loss or bone remodelling at sites where no change has occurred or as no change at sites where tissue has changed. The better the contrast correction method the less error it induces in the subtracted image. To analyse only the error induced by a contrast correction method, influences of projective geometry and physical noise from the camera and the analog-to-digital conversion were eliminated by simulating bone loss in digital copies of reference radiographic images.

The CDF method is based on the distribution of grey level values within the whole image or within a reference area which is approximately identical in two images. The results show that this method performed worse than the other two. From this study we can make the following two observations: 1) if loss is present in the reference region, the loss in all other regions will be underestimated and the remodelling overestimated, 2) if remodelling is present in the reference region, the remodelling in all other regions will be underestimated and the loss overestimated. The question is weather the CDF method would perform better in an *in vivo* study in which reference areas were chosen to normalise on. By our opinion no, because there are three problems associated with this method. First, the selected reference regions may contain actual changes in bone or tooth tissue, which can not be detected by visual inspection. Second, if an automatic procedure is to be used for image registration, contrast correction should precede registration. In this case it is hard to precisely outline two identical regions. Third, due to abundant changes, reference regions with adequate grey level distribution may not be found at all. The performance of the CDF contrast correction method is therefore closely related to the accuracy of defining reference regions and there always exists some uncertainty as one can not be sure whether the tissue of a reference region contains any loss or remodelling. To overcome this problem an image, probably smoothed, of a reference structure could be used to normalise on.[9]

The LSQA contrast correction method is performed by calculating the mapping function which is a regression curve through points representing grey levels of pixels having the same spatial co-ordinates in two images. There are two problems associated with this method. First, LSQA requires that the two images are previously aligned. We have excluded the influence of imperfect alignment by making copies of reference images. Second, actual changes between two radiographs deteriorate the outcome of normalisation. Results obtained for LSQA method are somewhat better than those obtained by the CDF method because the regression curve in the form of a quadratic polynomial is less sensitive to differences between two images than the CDF function. When applying the LSQA method in *in vivo* studies worse results are expected due to the problem of alignment.

The contrast correction method based on optical density thickness functions was briefly introduced by Vos et al.[7] The accuracy of ODTF recording significantly influences the performance of the ODTF method. Image noise affects the ODTF function but hopefully it can be minimised by averaging the wedge image. Image noise minimisation is correlated with the number of pixels included in averaging, hence, wide and long wedges and high image resolution should be used. It is also important that edges of an Al-wedge are properly defined. In this study they were defined manually, but if the edges are marked by adding a metal wire they could be detected automatically. Therefore, the ODTF method has the potential to be fully automated. The method yields the smallest amount of noise, consequently, less subtle tissue changes can be revealed.

In conclusion, it seems that less error is induced by a contrast correction method which uses a grey level mapping function based on the density distribution in a region representing a reference structure which is outside the lesion area. Although a reference structure which was used only in the ODTF method it could be used in the CDF and LSQA as well. The disadvantage is that if there is no wedge, because the x-ray was taken when DSR was not yet considered, the ODTF method can not be used, while CDF and LSQA will produce still meaningful results. On the other hand a method using a reference structure has the following advantages:

- it induces less error and is therefore more objective,
- it may deal with all grey level mismatches except those that are spatially dependent,
- CDF, ODTF, and LSQA functions may be successfully used to objectively enhance the contrast of conventional radiographic images and subtracted images,
- image quality can be evaluated and enhanced by shape analysis and synthesis of CDF, ODTF, and LSQA functions, respectively. This could lead to image quality standardisation,
- optimal shapes of CDF, ODTF, and LSQA functions might be defined for visual inspection of different diagnostic purposes (caries, periapical processes, and periodontal processes).

In addition to these advantages, the ODTF functions are further useful for estimating bone change volumes by densitometric image analysis.

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